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Kinematic adaptations to a variable stiffness shoe: mechanisms for reducing joint loading

[1,2]Boyer, K.A. [5,4]Federolf, P., [3,6]Lin, C, [4]Nigg, B.M., [2,1,3] Andriacchi, T.P.

[1]Mechanical Engineering, Stanford University
Stanford, CA, USA

[2]Bone and Joint Center, VA Palo Alto Health Care System
Palo Alto, CA, USA

[3] Department of Orthopedic Surgery, Stanford University Hospital,
Stanford University, Stanford, CA, USA

[4] Human Performance Lab, Faculty of Kinesiology,
University of Calgary
Calgary, Canada,

[5]Norwegian School of Sport Sciences
Oslo Norway

[6]Rehabilitation Institute of Chicago,
Spine & Sports Rehabilitation Center,
Chicago, IL

Corresponding author:
Katherine A. Boyer
kboyer@kin.umass.edu

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Introduction

Osteoarthritis (OA) is a major cause of disability in the aging population with the knee joint and it's medial compartment most often affected (Felson et al., 2000; Lawrence et al., 1989). Increases in the number of persons affected by the disease and escalating health care costs has led to increased interest in simple non-invasive interventions to modify both symptom and disease progression. One important marker and target for non-invasive intervention is the first peak of the external knee adduction moment in walking (Miyazaki et al., 2002). It is a surrogate marker of the relative load on the medial compartment (Schipplein and Andriacchi, 1991) and has been correlated with OA radiographic severity, rate of disease progression and severity of disease symptoms (Andriacchi et al., 2004; Andriacchi et al., 2009; Andriacchi and Mundermann, 2006; Astephen et al., 2007; Baliunas et al., 2002; Sharma et al., 1998).

One such intervention is a variable-stiffness shoe, a normal appearing athletic shoe with the midsole of the lateral aspect 50% stiffer than the medial side. A recent prospective randomized placebo controlled clinical trial found that pain was decrease by a clinically significant amount for patients wearing a variable stiffness shoe (Erhart et al., 2008; Erhart et al., 2010). This clinical benefit was attributed to the decrease in the external knee adduction moment, a change that has also been shown for a group of healthy subjects in this same shoe (Fisher et al., 2007). The variable stiffness shoe results are in contrast to studies with laterally wedged insoles that in both healthy and subjects with OA have had mixed results (Baker et al., 2007; Bennell et al., 2011; Fang et al., 2006; Kakihana et al., 2007; Kutzner et al., 2011; Toda and Tsukimura, 2004). The mechanism by which the external adduction moment is reduced in the VS shoe is not clear. Yet this information is important since the variable-stiffness shoe may reduce the adduction moment by a different mechanism than fixed devices such as the lateral wedge that aim to re-align the leg.

A previous study (Jenkyn et al., 2011) of the mechanism for the reduced adduction moment in the variable-stiffness shoe suggested that the shoe produced “coordinated dynamic changes” or changes in the position and motion of the lower limb segments to affect the center of pressure (COP) and the magnitude of the medio-lateral ground reaction force (GRF). These shifts in the COP and m-l GRF in turn reduced the frontal plane moment arm and resulted in a change in the external knee adduction moment. The adaptive movement response to the variable stiffness shoe has not been fully explored. It is therefore not known if there is a change in the motion or posture of the pelvis and leg in response to the variable stiffness shoe that would produce changes in the COP and GRF. This information is important for understanding how various load modifying devices actually reduce the adduction moment.

Identifying a kinematic response to an intervention is challenging. This challenge can be due to the combination of inherent biologic variability in gait patterns, where between subject differences are larger than the within subject differences (Stacoff et al., 2001) and/or because the expected kinematic changes particularly in the frontal plane are small (Jenkyn et al., 2011; Nigg et al., 2003; Stacoff et al., 2000). It is also challenging to select variables that describe both changes in joint posture and joint or segmental motion that may be present and important to describe the effect of an intervention. Higher order statistical methods such as a principal component analysis (PCA) offer methods that are data driven and potentially more sensitive than traditional discrete variable analysis methods to detect small systematic differences in joint postures and motions (Daffertshofer et al., 2004; Nigg, 2010). The aim of this study was to test the hypothesis that:

a) there are differences in the frontal plane joint positions and motions and

b) these are correlated with differences in the medio-lateral ground reaction force and center of pressure for the variable-stiffness shoe compared to a constant stiffness control shoe in a group of healthy adults.

Methods

Eleven healthy adults performed five walking trials at a self-selected speed in both a constant-stiffness control shoe and a variable-stiffness shoe. The shoes were both generic athletic designs with nylon/leather upper material and a cushioned sole. The sole of the variable stiffness shoe is made of ethylene vinyl acetate (EVA). The sole was 1.3 to 1.5 times stiffer on the lateral side of the shoe compared to the medial side. Asker C durometer values for the medial sole were 55+/- 2 while values for the lateral sole were 70-76 +/- 2. The control- shoe was similar in design but with a uniform sole stiffness comparable in stiffness to the medial sole stiffness of the intervention shoe.

The protocol was approved by the Stanford internal review board and informed written consents were obtained. Marker-based motion data was collected with a 10 camera optoelectronic system (Qualisys AB, Gothenburg, Sweden) sampling at 120Hz. Marker trajectories were labeled using the Qualysis Track Manager software and exported for further processing in custom MatLab © software (MATLAB R2010b MathWorks inc 1984-2010). A multi component force plate (Bertec Corporation, Columbus, Ohio) embedded in the floor was used to capture ground reaction force data synchronously at 1200Hz.

Markers were placed on the anterior and posterior superior iliac spines, the iliac crests, on the lateral and medial femoral condyles, on the medial and lateral aspect of the malleolus, medial and lateral aspect of the calcaneus, and at the base of the fifth metatarsal. Additionally clusters of nine and seven reflective markers were distributed on the thigh and shank respectively. Subject wore split-leg running shorts for the testing and all markers were placed on the skin using double sided tape. An anatomical standing reference trial was recorded with the subject standing still to create the anatomical reference frames for each limb segment (foot, tibial, femur and pelvis) using the markers placed on the bony landmarks. The hip joint center was defined using a functional approach outline by Halvorsen et al., (1999). A previously described point-cluster technique (Andriacchi et al., 1998), which uses a redundant set of markers on the thigh and shank was used to estimate the rotations of the tibia with respect to the femur. The motion of the knee was determined by relating the motion of the marker clusters to the anatomical coordinate systems. Details of the axes orientations of the femoral and tibial anatomic coordinate systems have been described previously (Andracchi 2003, 2005, Dyrby and Andriacchi 2004, Scanlan et al., 2010). The pelvic and foot coordinate systems and corresponding hip and ankle joint complex coordinate systems followed the ISB recommendations (Wu et al., 2002). Segment kinematics were filtered at 6Hz using an 8^{th} order Butterworth filter. The stance phase of walking was identified using the vertical ground reaction force data. A threshold of 5% of bodyweight was used to identify heel-strike and toe-off. An inverse dynamics approach was used to calculate the external knee adduction moment. The moments were normalized to % bodyweight X height. Walking speed for each trial was determined using the horizontal velocity of the markers placed on the posterior superior iliac spines.

Data analysis

A PCA was used to identify correlated changes within the variability of the 17 variables of interest (Daffertshofer et al., 2004; Nigg, 2010), These were the three dimensional joint kinematic waveforms of ankle, knee, hip and pelvis, the ground reaction forces (GRF) and the center of pressure (COP) positions. Principal components that quantified changes due to the shoe condition were identified and their combined effects on the 17 variables were analyzed.

The kinematic and force-plate data was prepared using two normalization steps: 1) All waveforms were interpolated such that 100 points represented the stance phase and the average for the five trials for each subject and shoe was calculated for each variable 2) the mean over the stance phase was subtracted from each variable and the amplitude divided by the standard deviation. This step was necessary to standardize all the variables as the PCA is scale-sensitive. The resultant normalized waveforms of all variables for each subject and shoe were appended together into one column trial vector with 1700 vector components (variables x time-points). Each subject's data can be visualized as a single point in a high dimensional data space. If this subject walks in a different shoe and with different gait mechanics then the position of this point in space will move.

Data for all subjects and shoe were assembled in a single matrix (DATA) for the analysis. The covariance matrix was computed for the matrix DATA and then an eigen decomposition of the covariance matrix was performed see von Tscharner (2002) for a detailed descriptions of the mathematics). The eigenvectors or principal components (PC_i) represent the primary directions of variance in the data. The eigenvalues (EV_i) (expressed as a percentage of the sum of all eigenvalues) indicate the variance (%) in the data set explained by a particular PC_i. There are 22 trials (11 subject, 2 shoe conditions) and thus 21 PC_i contain all information regarding directions of data variation. Calculations were implemented in Mathematica 7.0 (Wolfram Research Inc 1998-2009).

Each PC_i represents a characteristic manner, related to any or all of the variables and time-points, of how individual trials deviate from the average gait pattern. A PC vector that has more than one non-zero component represents variation in more than one variable or time-point and these variations are correlated. Each data vector can then be described in terms of the mean of all trial vectors and a weighted linear combination of the PC_i. The weights for each trial are determined by projecting the normalized trial vector onto the PC_i. A difference in the weighting factors between shoes for any of the PCs would indicate that along that direction the shoes are different.

A combination of a paired student's t-test and Cohen's d statistic (effect size for change) was used to determine if there were significant differences in weighting factors for a PC_i indicating a systematic change in the waveforms due to the different shoe conditions.. A linear combination (using the EV_i as scaling factors) of those PC_i with $d > 0.6$ and $p< 0.1$ defined a discriminant vector incorporating the systematic differences in gait characteristics that occurred with the change of shoe. The subjects' trial vectors were projected onto this discriminant vector and differences in the weights between the two shoes were tested for again. The main non-zero components of discriminant vector can be related back to the original kinematic /GRF variables for the interpretation (Figure.1). Additionally, the discriminant vector (for each shoe) was added the mean of the trial vectors and normalization steps retraced to allow a visualization of the differences in gait characteristics between shoes for interpretation of the direction of change (Figure. 1,2, and 4). An illustration of this procedure is provided in Figure 1. Differences in gait variables were amplified for the visualization by multiplying the weight factors with an amplification factor of 10.

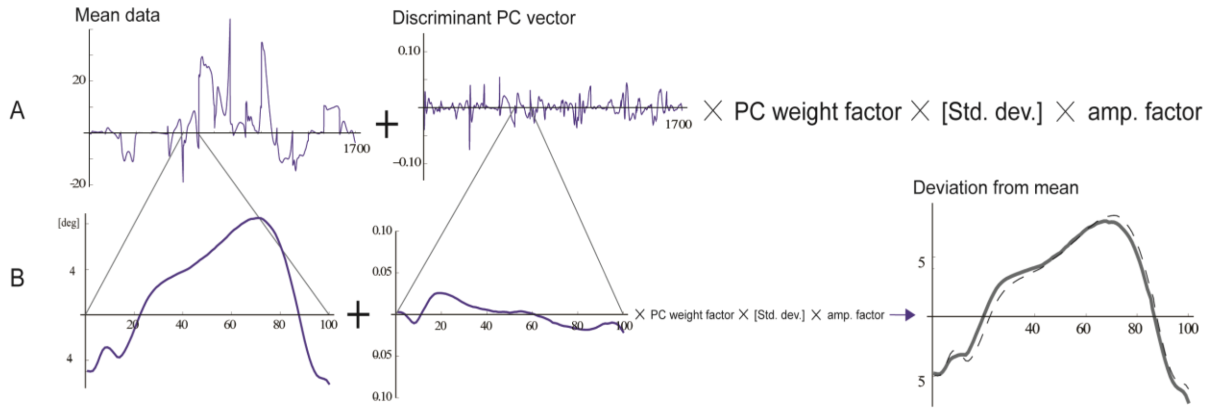

Fig 1. A) The mean data vector for both the variable stiffness and control shoe plotted as a time series. The GRF, COP and kinematic waveforms are stacked end on end to create a single mean vector (1700 components long) for the analysis procedure. The disciminant PC vector, a linear combination of those PC_i that were significantly different between shoe conditions is also plotted as a time-series. To visualize the differences in gait characteristics identified by the analysis the normalization step is retraced and the discriminant vector times the mean weighting factor for each shoe condition and an amplification factor is added to the mean of the trial vectors. We can focus on small segments of the larger data vectors to interpret the results in terms of biomechanical variables.

B). For example the deviations from the mean ankle in/eversion for one shoe condition are illustrated by the thick line in the figure at the far right. Where the thick dark line is the same as the line dashed line (the mean data vector) there is no systematic effect of a shoe condition. Where the thick line differs from the dashed line, we can say there was a systematic effect of the shoe intervention for this variable at these time-points. To illustrate the results of this study the deviations from the mean data that were evident using the amplification factor of 10 have been plotted in Fig 2.

Results

The projection of the subjects' trial vectors onto the linear combination of PC_4, PC_5, PC_9, and PC_{10} was significantly different between the two shoe conditions (*d* =1.77; $p < 0.001$). The differences in gait patterns between the control and variable-stiffness shoe were interpreted by plotting the portion of the discriminant vector plus the mean trial vector corresponding to each variable (Figure 2.) This indicated for the variable stiffness intervention shoe there was a) a change in the shape of the ankle plantar dorsi-flexion waveform with greater dorsi-flexion between 15 and 60% stance but a smaller peak dorsi-flexion angle in terminal stance. b) a change in the motion of the ankle in inversion-eversion with an increase in the eversion and eversion velocity between 15 and 45 % stance but a smaller peak eversion angle in the terminal stance phase. c) slight increases in the knee abduction angle but also adduction angle during the loading phase of stance (5-25%) d) less femur internal rotation with respect to the tibia e) less hip adduction in the first half of stance and 6) smaller pelvic obliquity in the first 40% of stance (i.e. a more level pelvis).

In addition there was a smaller medial and lateral GRF along with a more neutral COP position in the first 30% of stance. A slight increase in the peak posterior GRF and an earlier timing of the peak vertical GRF were also found. There was an 8.7 % +/- 3.49% decrease in the first peak external knee adduction moment for the VS shoe compared to the control shoe (Figure 2). The timing of the decrease in the first peak of the adduction moment was consistent with the kinematic changes illustrated in Figure 2 & 3. The mean walking speed was not significantly different between the two shoe conditions (VS 1.36 +/- 0.03 m/s; control 1.34 +/- 0.06 m/s)

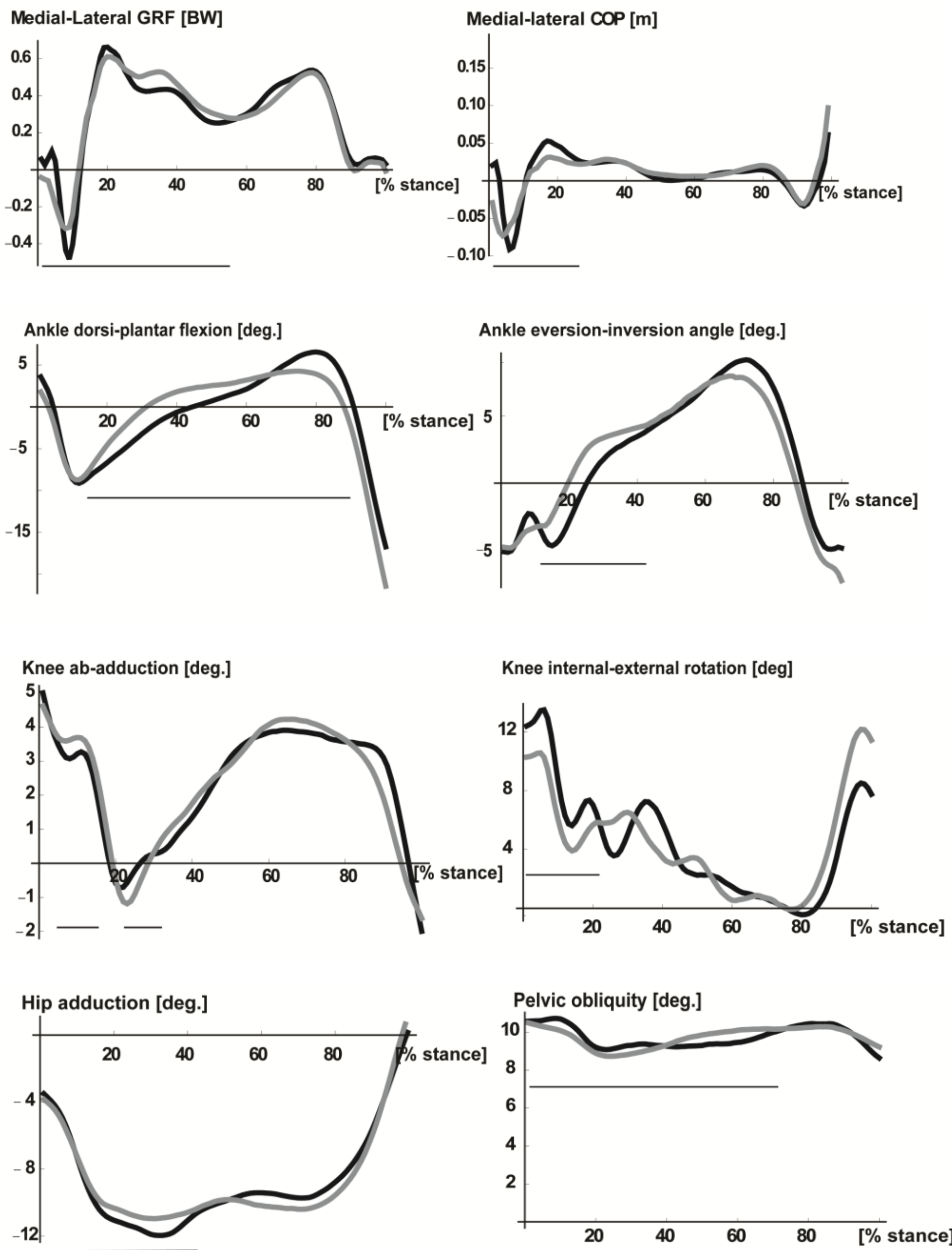

Figure 2: The mean deviations in joint angles, COP and GRF from the mean kinematic waveforms for the variable stiffness (gray) and control shoes (black). The waveforms shown are those for which there were corresponding non-zero values for at least 5% of stance along the discriminate vector. These deviations from the mean gait, described by the discriminate vector, have been amplified by a factor of 10 for visualization. The horizontal bars indicate the time frame where the disciminant vector was different from zero. The timing of the kinematic difference between shoes corresponds with the changes in the external knee adduction moment (Figure 3).

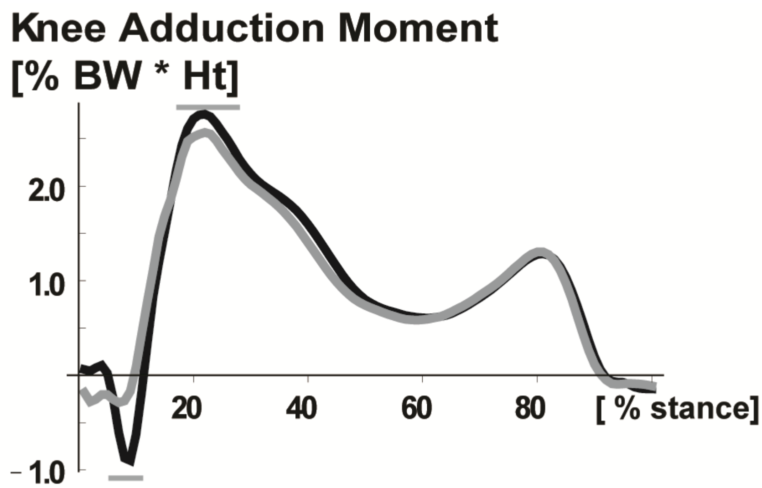

Figure 3: The mean external knee adduction moment for the control shoe (black) and VS shoe (gray). The horizontal bars indicate the region of the stance phase where there are differences between the control and VS shoe. Difference between the control and VS shoe occur within approximately the first 30% of stance.

The four principal components, PC_4, PC_5, PC_9, and PC_{10},that made up the discriminant vector were those components that had both large effect sizes(Cohen's *d* ≥0.6) and small p-values (< 0.1) for a paired students t-test for the difference between the control and intervention shoe. (Table 1). Together these PC_i explained 25.6 % of the variance in the data, which can be attributed to the different shoes.

Table 1: For the first 16 PC's the relative variability explained (EV_i), Cohen's *d* comparing the intervention and the control shoe and exact p-values for those less than p=0.1.

PC_i	1	2	3	4	5	6	7	8
EV_i [%]	19.1	14.3	12.9	**10.8**	**8.8**	8.0	5.0	4.0
Cohen's d	0.30	0.13	0.25	**-1.30**	**0.60**	-0.46	0.46	0.17
p-value				0.001	0.07			
PC_i	9	10	11	12	13	14	15	16
EV_i [%]	**3.3**	**2.7**	2.2	1.7	1.5	1.2	0.9	0.8
Cohen's d	**-2.57**	**0.72**	-0.27	-0.24	-0.31	-0.02	0.13	0.005
p-value	**0.000**	**0.04**						

Discussion

The aim of this study was to identify the dynamic response to the variable stiffness shoe that would contribute to changes in the external knee adduction moment. Using a PCA analysis of trial vectors consisting of all joint kinematics, GRF and COP data we identified important differences between the variable-stiffness and control shoe in the ankle eversion, knee abduction and adduction, hip adduction and pelvic obliquity angles that would change the posture of the leg in the frontal plane, in addition less knee internal rotation (femur with respect to tibia) was found. Together these changes in joint kinematics can be interpreted as a more vertical leg and pelvis position with the pelvis positioned directly over the weight bearing leg for the intervention shoe as compared to the constant stiffness control shoe (Figure 4). The discriminant vector differentiating the two shoe conditions also indicated there was a reduced excursion of the COP and peak medial and lateral GRFs for the variable stiffness compared to the control shoe. Our results are in agreement with the suggestion that it is a dynamic adaptation (Jenkyn et al., 2011) or a change in the relative motion of the segments (not realignment of the skeleton or constant offset in joint positions), to the variable-stiffness shoe that contributes to the change in joint loading.

The timing of the differences in kinematics and GRF would be expected contribute to changes in the resultant knee adduction moment (Figure 2 & 3). Thus, the results of this study, by providing a biomechanical explanation for the change in GRF and COP, indirectly explain the change in the knee adduction moment with the variable-stiffness shoe. A change in the line of action of the ground reaction force in combination with a shift in the m-l COP will contribute to a change in the relative distance from the center of the joint to the force vector (i.e. the moment arm) . Changes in the m-l GRF and m-l COP have previously been shown to explain 50% of the variance in the change of lever arm at the knee (Jenkyn et al., 2011). The lever arm in that study was calculated as the projection of the vector from the COP to the knee joint center onto the line of action of the ground reaction force. The adaptive response to the variable-stiffness shoe was subtle and this was expected and a motivating factor for the use of the PC analysis. Based on previous work a change in the frontal plane lever arm (distance from GRF vector to knee joint center) on the order of 2-4% or 2 to 4 mm would be expected in conjunction with changes in the knee adduction moment (average 8.3% % BW * Ht) found in our study with healthy individuals (Fantini Pagani et al., 2011; Jenkyn et al., 2011).

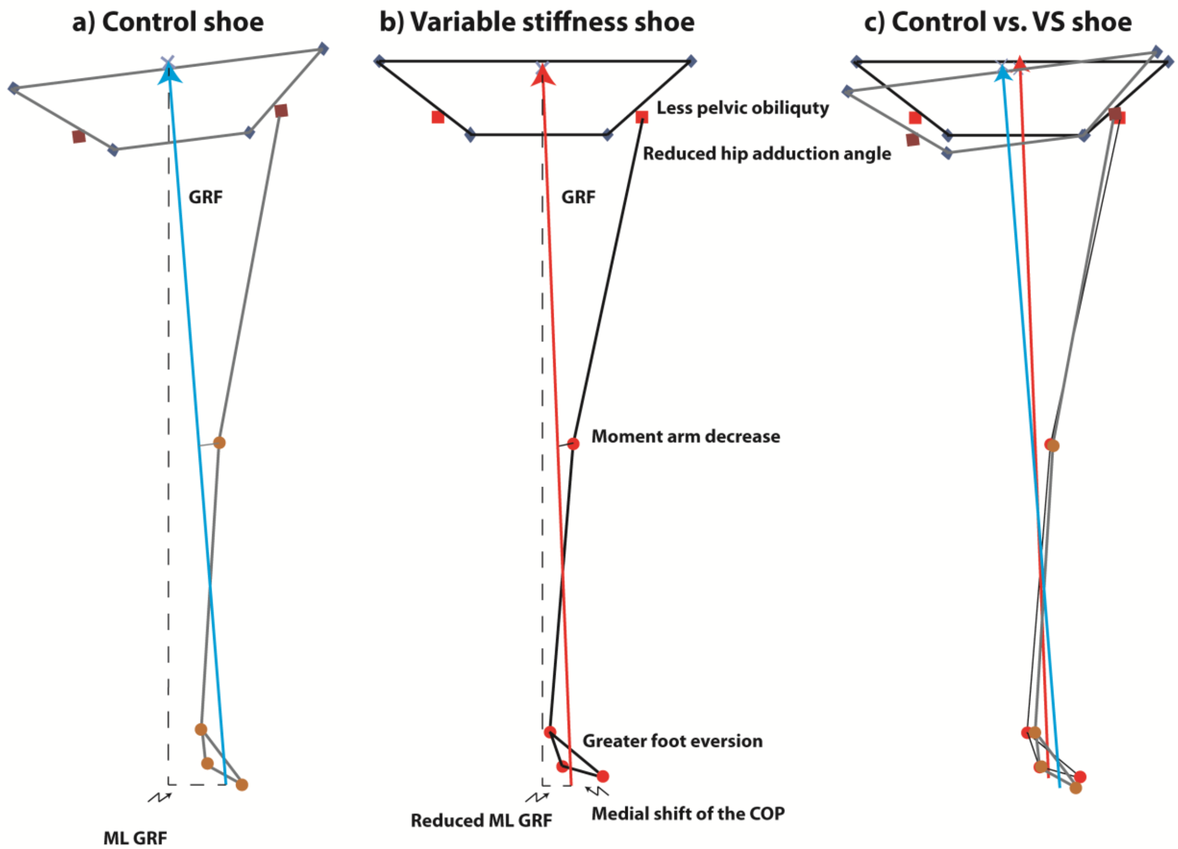

Figure 4: An illustrative comparison of the frontal plane joint kinematics in the control (a) and variable stiffness shoe (b) and the relationship with the ground reaction force, body center of mass, frontal plane moment arm and center of pressure under the foot (COP). The adaptive response to the variable stiffness shoe was an increase in ankle eversion, less hip adduction and more level pelvis. As illustrated this would result in a more vertical leg and pelvis position and a change in the direction of the GRF vector.

A comparison of the dynamic changes associated with the variable stiffness shoe to kinematic changes in fixed interventions such as lateral wedge shoes might help to explain the difference in the clinical outcome (Erhart et al., 2010) with this shoe relative to the inconsistent results seen with wedged interventions (Kakihana et al., 2007; Toda and Tsukimura, 2004). The primary difference between response to lateral wedge interventions and the variable stiffness shoe is the direction of shift in the COP in combination with the change in the m-l GRF (Fantini Pagani et al., 2011; Kakihana et al., 2007). A lateral shift in the COP and in some instances an increase in the m-l GRF has been reported for the lateral wedge interventions (Fantini Pagani et al., 2011; Hinman et al., 2012; Kakihana et al., 2007). The kinematic changes that were associated with these changes in the reaction forces are a slight increase in the abduction angle and greater ankle eversion. This study found, in agreement with a previous study (Jenkyn et al., 2011), for the variable stiffness shoe there is a medial shift in the COP and a reduction in the both the peak medial and lateral GRFs.

The kinematic changes associated with this shoe were greater ankle eversion in early stance and knee abduction angles, similar to the lateral wedge. In addition, a reduced hip adduction angle and pelvic obliquity angle in the first half of stance were found with the variable stiffness shoe. This is in contrast to one previous report for lateral wedge shoes, where an increase in the hip adduction angle was reported (Hinman et al., 2012). This difference is a likely contribution factor, along with the pelvic angle, to the differential response with respect to the COP and m-l GRF change for the variable stiffness versus the lateral wedge interventions.

The results of this study provide a unique insight into the interaction of segment movements that characterize the response to the variable stiffness shoe. In addition to difference in the joint angles at or around the time of the 1st peak knee adduction moment, the PCA identified differences in the rate of eversion motion at the ankle, subtle but systematic differences in the timing of the peak knee abduction angle and also highlighted differences in the shoe response between early and late stance. It is important to note that the subtle changes in movement were only visible because the PCA method permits amplification of the statistically meaningful components of movement. The application of the PCA method also has some advantages over traditional analysis methods regarding study sample size needed to detect significant differences, a significant response was found in this study with a sample size of only eleven subjects. This has substantial advantages for efficient development and testing of novel mechanical interventions for musculoskeletal injuries.

This study used a healthy subject population although we expect that the response will be the similar in an OA population given the similarities in the changes between the two populations in the knee adduction moment, COP and medial-lateral GRF. However, in the most severe OA patients, the effects of severe leg mal-alignment and/or muscle weakness may contribute to greater variation in the dynamic response to the variable shoe. This analysis identified those changes in gait that were systematic between the two shoe conditions; additional individual changes in gait may also be present that could increase the magnitude of the response. Moments were not included in the PC analysis; due to the limited sample size we chose to limit the number of variables to be less than the number of PC that could describe the variation in data. In addition, our primary interest were those kinematic adaptations that may be connected with the COP and m-l GRF changes as these had previously been identified as the key components the lead to changes in the adduction moment in an OA populations.

Conclusions

This study has demonstrated the mechanism that explains the changes in the GRF and COP in response to the VS shoe in healthy adults. Using a PCA this study found that 25% of the variance in the dataset could be attributed to the change in shoe condition. This adaptive response is subtle changes in the frontal plane motion of the ankle, hip and pelvis that are connected with changes in medial lateral force and COP. The GRF and COP have previous been correlated with the decreased external knee adduction moment (Jenkyn et al., 2011), a clinically important variable for medial knee OA related to both symptomatic and radiographic disease status and progression (Miyazaki et al., 2002). Understanding the dynamic response to a non-invasive intervention such as the variable-stiffness shoe is a first research step towards customization and optimization of interventions for patients.

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Conflict of interest statement

All authors deny any financial or personal relationship with people or organization that could potentially bias this work.

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